EVIDENCE REPORT:

RISK OF INJURY DUE TO DYNAMIC LOADS

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PRD RISK TITLE: Risk of Injury Due to Dynamic Loads

DESCRIPTION: Given the range of anticipated dynamic loads transferred to the crew via the vehicle, there is a possibility of loss of crew or crew injury during launch, abort and landing.

1.0 EXECUTIVE SUMMARY

During spaceflight, crewmembers are at risk of injury due to dynamic load exposure. Dynamic loads are transient loads (≤500ms) which are most likely to occur during launch, pad or launch abort, and landing.

Several extrinsic factors affect the risk of injury including: vehicle dynamic profile, the design of the seat and restraint system, as well as the spacesuit and helmet. Because each vehicle can have different launch, abort and landing dynamics, the risk of injury is greatly influenced by the vehicle design. Vehicles which minimize crew exposure to dynamic loads will be inherently safer than vehicles which have higher dynamic loads. The seat and restraint designs may either increase or mitigate risk of injury depending on how effective they are at minimizing movement of the human body relative to the seat and other body regions. Finally, the spacesuit and helmet may contribute to the risk of injury if the design is not configured for occupant protection during dynamic loads. For instance the suit can hinder the effectiveness of the restraints on the crewmember thus magnifying dynamic exposure. Rigid elements of the suit can induce point loading, while the mass of the helmet poses a risk of injury such as blunt impact or neck overloading if not properly supported.

In addition to these extrinsic factors, there are additional intrinsic factors of the crew that can contribute to the risk of injury, such as age, gender, anthropometry, and deconditioning due to spaceflight. Age has been shown to be a risk factor in other analogous environments such as automobile collisions. Gender can also influence injury risk, as body strength and geometry can differ between men and women. Anthropometry has been found to have an effect on injury risk since loads may not be proportional to the difference in anatomical structure and strength. Finally, spaceflight deconditioning has been shown to degenerate the structural and tissue responses in the musculoskeletal system which imply the crewmember may have a lower tolerance to dynamic loads.

To assess the risk of injury, there are multiple methods available, although each has advantages and disadvantages. The methods can be divided into 3 categories: humans, human surrogates, and numerical models. Although human data seem to be the ideal solution for assessing injury risk, there are several drawbacks. Human volunteer testing is limited to sub-injurious levels but allows subjective feedback. Post-mortem human subjects (PMHS) can be tested at injurious levels, but cannot be used to investigate living tissue responses to trauma and do not include active muscle tone. Human exposure data contains cases of living human injury, but do not allow for prospective investigations of injury mechanisms. Human surrogates include Anthropomorphic Test Devices (ATD) and animal models. ATDs are not biofidelic in all instances and are not able to predict injury in all conditions; however, they are easily tested and have reproducible data. Animal models allow prospective testing of living tissue, but are not anatomically identical to humans. In addition, numerical models are available to assess injury risk. Dynamic response models are simple, but are limited in their injury prediction capabilities. ATD Finite Element (FE) models have similar limitations as the physical ATDs. Human FE models have great potential for allowing injury predictions; however, currently they are not validated in all necessary conditions. Finally, regardless of the method used to assess injury risk, adequate criteria for assessing low risk of injury (<5%) are needed.

Given this evidence, multiple knowledge gaps still exist in our understanding of the risk of injury to dynamic loads. These gaps include: the effect of various body orientations on injury risk during spaceflight; the effect of suit, seat and restraint designs on injury risk; the effects of age, gender and anthropometry on injury risk; the effects of spaceflight deconditioning on injury risk; criteria to adequately assess low risks of injury; and adequate methods for assessing injury risk. These knowledge gaps highlight area of needed research to assist in mitigating the risk.

2.0 INTRODUCTION

2.1 Context

The nature of spaceflight dictates an extreme amount of kinetic energy to reach space, and effective systems to dissipate this energy during the return to earth. While most of this energy is dissipated or absorbed by the vehicle, some amount of kinetic energy will be transmitted to the occupants aboard the spacecraft. This energy, if not properly managed, may cause injury to the crewmembers.

With the retirement of the Space Shuttle, the National Aeronautics and Space Administration (NASA) is involved in the development of several vehicles with differing landing modes and conditions. Any injuries that may occur as a result of excessive energy imparted to the vehicle’s passengers may impair or prevent a crew-member from unassisted evacuation of the spaceflight vehicle after landing. Unfortunately, the current NASA standards and requirements do not adequately address injury risk from many key factors. This was highlighted in the *Columbia* Crew Survival Investigation Report, which cited inadequate upper body restraint and protection as a potential lethal event. The report recommended that future spacecraft suits and seat restraints should use state-of-the-art technology in an integrated solution to minimize crew injury and maximize crew survival in off-nominal acceleration environments as titled L2-4/L3-4. It also recommended future vehicles should incorporate conformal helmets and neck restraint designs similar to those used in professional auto racing outlined in L2-7 of the report [1].

Development of Agency-level human health and performance standards appropriate to occupant protection from dynamic loads as well as development of the method(s) of meeting those standards in the design, development, and operation of mission systems would allow vehicle designers to mitigate the risk of injury in their designs, reducing the likelihood of crew injury or Loss of Crew (LOC).

2.2 Scope of Occupant Protection

Given the range of anticipated dynamic loads transferred to the crew via the vehicle, there is a possibility of loss of crew or crew injury during launch, abort, and landing. This report provides evidence of this risk based on two major groups of contributing factors – extrinsic and intrinsic. These factors influence the dynamic loads transmitted to the body, human inertial response and human tolerance/limits during dynamics.

Currently NASA's Occupant Protection is tasked with mitigating risk to crews due to inertial responses of occupants to transient accelerations. Transient accelerations are defined as accelerations lasting for less than 500ms. Included are any elements that a crewmember may contact during dynamic phases of flight: the seat, restraint system, spacesuit, helmet and any component the occupant could contact during flail. Precluded are the supporting structural elements of the vehicle such as the walls, floor, struts, etc. It is presumed that these structural elements will remain intact maintaining the occupant volume during all phases of flight and will not impinge upon the crew during structural loading. NASA occupant protection standards and practices will be applied within this scope.

2.3 Definition of Injury

Although a number of injuries are possible during dynamic phases of flight, the risk of injury that is addressed in this report are musculoskeletal and soft tissue injuries due to dynamic loads. Other controls are in place, which address other aspects of injury such as burns, inhalation, and decompression sickness.

In addition, injuries are defined further by the Operational Relevant Injury Scale (ORIS) [2]. The ORIS is a NASA developed injury scale, which accounts not only for injury severity, but also significance. Three elements are used to determine a composite score: severity, vehicle self-egress capability, and crewmember return to flight status. Injury severity is based on the Abbreviated Injury Scale (AIS) severity classification [3]. Self-egress ability is a measure of a crewmember's capacity for autonomously exiting the vehicle after landing. Since crewmembers may be required to egress the vehicle immediately, injuries could have operational consequences that are not captured in the AIS severity score. Finally, return to flight status is a measure of long-term consequences, which may be unique to NASA. For example, an injury that is classified as having no long term consequence for an average civilian could possibly disqualify an Astronaut from future flights.

Another important distinction when defining injury relates to possible spaceflight-induced conditions. Motion sickness or muscular deconditioning due to microgravity exposure may prevent an uninjured crewmember from self-egressing. This degradation in performance is not included in the definition of injury for our purposes; however, any increase in injury risk due to these factors will be addressed below.

2.4 Definition of Dynamic Loading

Dynamic loading is defined as the acceleration of a mass for transient periods (≤500 ms) [4]. Profile accelerations may be characterized by the onset, magnitude, duration and offset acceleration as shown in Figure 2-1 referred to as a trapezoid pulse [5, 6].

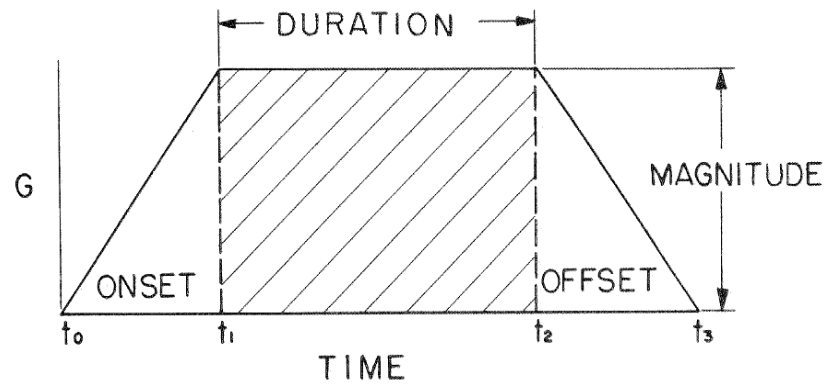
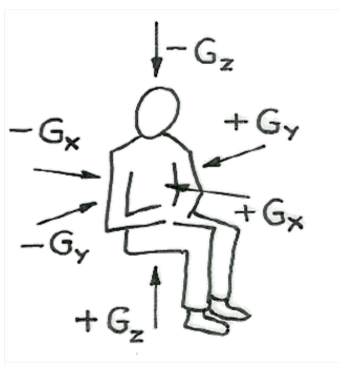


Figure 2-1: Acceleration Profile as Trapezoidal Pulse [6]

The trapezoidal pulse was often used in literature as a basis of comparison to describe input accelerations applied to the body, or vector direction as shown in Figure 2-2.



Upward accel. (positive G)	$+G_z$	Eyeballs down
Downward (negative G)	$-G_z$	Eyeballs up
Backward accel.	$+G_x$	Eyeballs out
Forward accel.	$-G_x$	Eyeballs in
Acceleration to right	$+G_y$	Eyeballs left
Acceleration to left	$-G_y$	Eyeballs right

Figure 2-2: Direction of Acceleration Relative to Body [7]

It is important to note that the human acceleration response is generally different than the acceleration input. Figure 2-3 illustrates input sled acceleration "A" which resembles a trapezoidal pulse, computed from displacement and velocity. Though the sled acceleration looks like a trapezoid, the measured acceleration on the subject, "B," looks like a half-sine wave. Furthermore, the human acceleration response is also dependent on the direction of the acceleration input. Therefore, fitting of the trapezoidal acceleration-time histories to assess human response for complex multi-directional landing would be inadequate to predict risk of injury. To identify human tolerance levels, both human response and the dynamic loading measures are required.

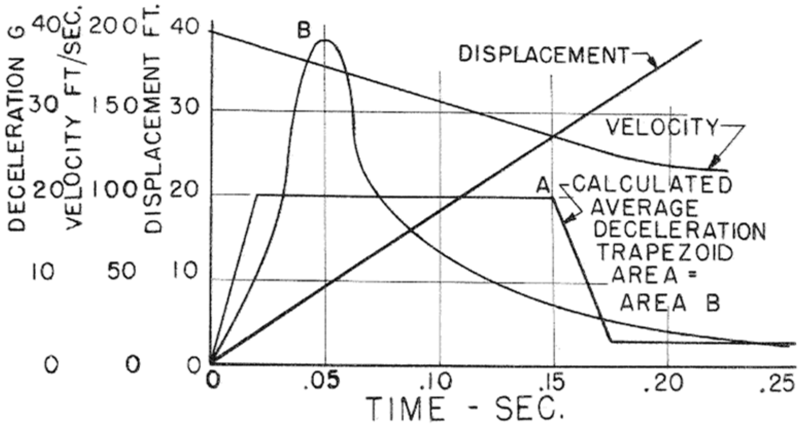


Figure 2-3: Acceleration Input to the Sled versus Acceleration of the Body [6]

2.5 Tissue Response Due To Dynamic Loading

Occupant Protection focuses on musculoskeletal injury based on biomechanics. Biomechanics which is defined as, "the science that examines forces acting upon and within a biological structure and the effects produced by such forces [8]." As a force makes contact with the body, it applies pressure over a given surface area, which is referred to as tissue stress. In turn, the tissue deforms resulting in tissue strain (deformation). The stress/strain relationship may be characterized by dimension (uniaxial, biaxial or triaxial) and direction (tension, compression, bending or torsion).

Every tissue in the body has a unique stress/strain characterization similar to Figure 2-4 [9]. If stress/strain is applied within the elastic range, the material is able to resume its original shape once the load is removed. If an exposure is applied beyond the tissue's yield point (outside the elastic region), the structure is compromised and will not return to its original shape. If the tissue is loaded to failure point (ultimate stress/strain) the tissue will tear or break. Compromised or damaged tissue would result in crew injury, which could lead to loss of mission and/or crew.

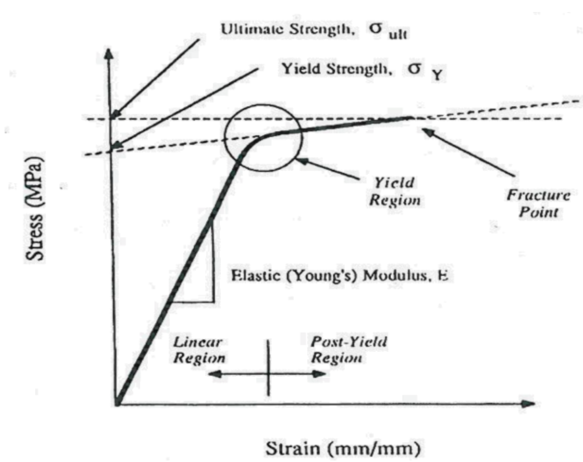


Figure 2-4: Stress/Strain Properties of Bone in Tension [9]

3.0 EVIDENCE

Evidence of injury due to dynamic loading is separated into two major factors; extrinsic and intrinsic (Figure 3-1). Extrinsic factors are hardware dynamics that include vehicle acceleration, suit/helmet design, seat and restraint system. Intrinsic factors incorporate physiological parameters that influence injury tolerance such as: age, gender, anthropometrics and spaceflight deconditioning.

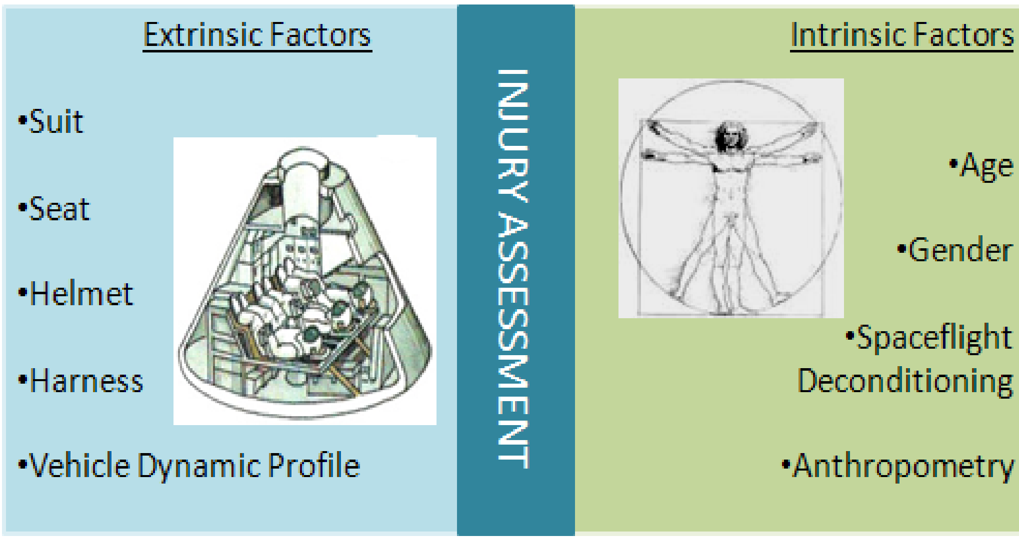


Figure 3-1: Risk of Injury Factors

3.1 Extrinsic Injury Risk Factors

3.1.1 Vehicle Dynamic Profile

3.1.1.1 Introduction

The vehicle dynamic profile is related to the design of the vehicle, which is driven by the space mission. With the recent retirement of the Space Shuttle, NASA is involved with the development of several different vehicles. Figure 3-2 illustrates various designs for future space vehicles, including capsule and lifting body designs, which are under consideration by NASA. Each vehicle will have unique dynamics depending on the design of the launch, abort, reentry and landing systems; however, regardless of the vehicle, the nature of spaceflight automatically exposes crew to dynamic loading in various directions. Although each vehicle may have different inherent injury risk, it is known that dynamic loading can induce injury and the injury threshold is dependent on the direction of acceleration. Evidence of injury due to dynamic load in different environments is outlined further in this report.

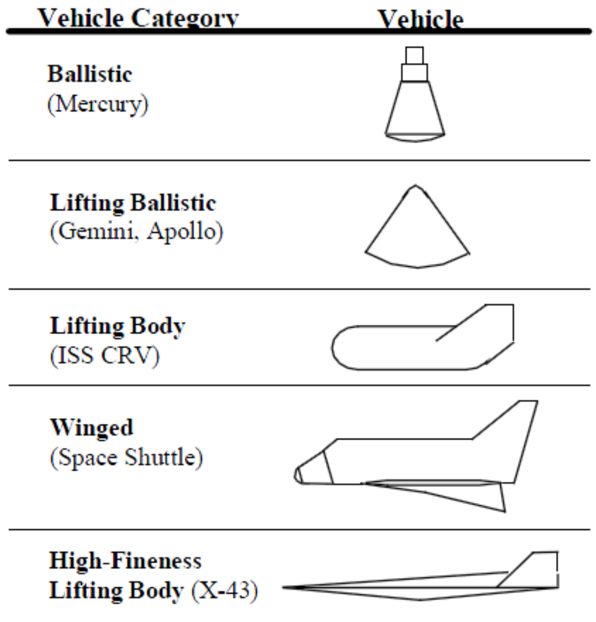


Figure 3-2: Vehicle Shapes [10]

3.1.1.2 Automotive Evidence

In 2009 in the US, 33,808 fatalities and 2.2 million injuries were reported from motor vehicle crashes. Of the fatalities, 10,591 were speeding related, and 4,885 occurred on roads with a posted speed limit of 55 mph or higher [11]. Markogiannakis, et al. reported anatomical regions of sustained injuries in motor vehicle cases in which 50% of the occupants sustained head injuries, while only 10% sustained spinal or pelvic injury (Table 3-1) [12].

Table 3-1: Distribution of Motor Vehicle Injuries by body region [12]

Body Region	*N*	*Percent*
Head	365	50%
Thorax	222	30.4%
Abdomen	104	14.2%
Spinal Cord	70	9.6%
Pelvis	68	9.3%
Upper and Lower Extremity	265	36.3%
Total	**730**	**100%**

In addition to anatomical injury regions the overall risk of injury and severity to a specific anatomical region may be determined due to tissue response from dynamic loads. For instance, the Head Injury Criteria (HIC) is used in the automotive industry to assess injury risk related to head acceleration. Figure

3-3 shows the National Highway Traffic Administration injury risk curves for HIC 15 [13]. Further studies found improved harness restraint systems increased tolerance levels for the onset acceleration rate [5, 14].

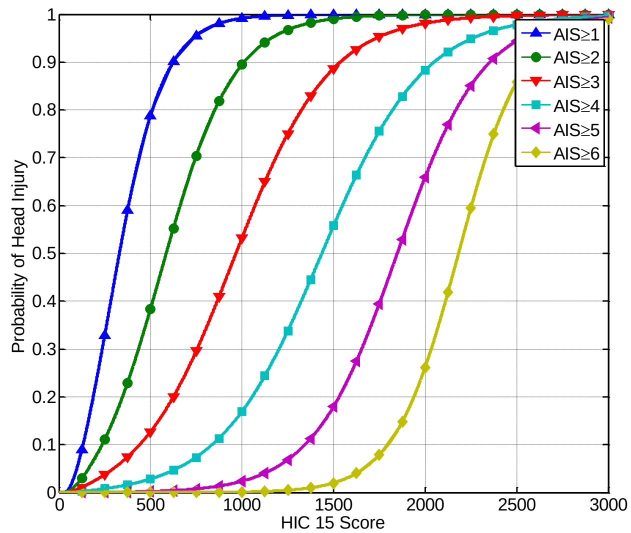


Figure 3-3: HIC 15 Injury Risk Functions [13]

3.1.1.3 Automotive Racing Evidence

Automotive racing exposes drivers to extreme vehicle dynamics as seen in Figure 3-4 [15]. Using the few injuries that occurred (41 out of 4015), the risk of head injury was determined related to HIC 15 (Figure 3-5). In this case, the risk of head injury was found to be significantly lower than previous research, suggesting that the seat and helmet played an important role in reducing the injury risk.

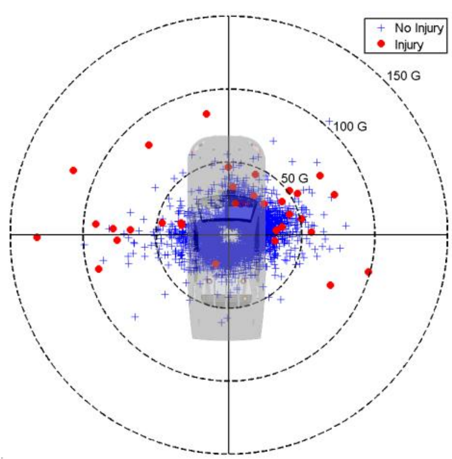


Figure 3-4: NASCAR Injury Distribution [15]

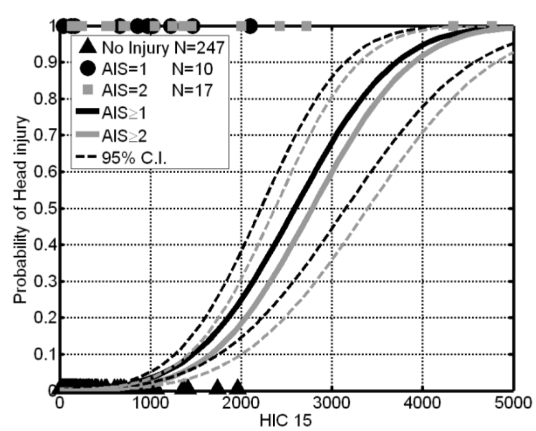


Figure 3-5: NASCAR Head Injury Risk [15]

3.1.1.4 Military Aircraft Evidence

The United States Air Force interest in biomechanics aims to protect pilots during off-nominal mission events: such as seat ejection, high G maneuvers and failure of landing systems. Operationally, pilot ejections have been shown to induce thoracolumbar spinal injury. The rates of injury operationally vary by aircraft but can be as high as 35%, as in the F-4 (see Figure 3-6). [16, 17]

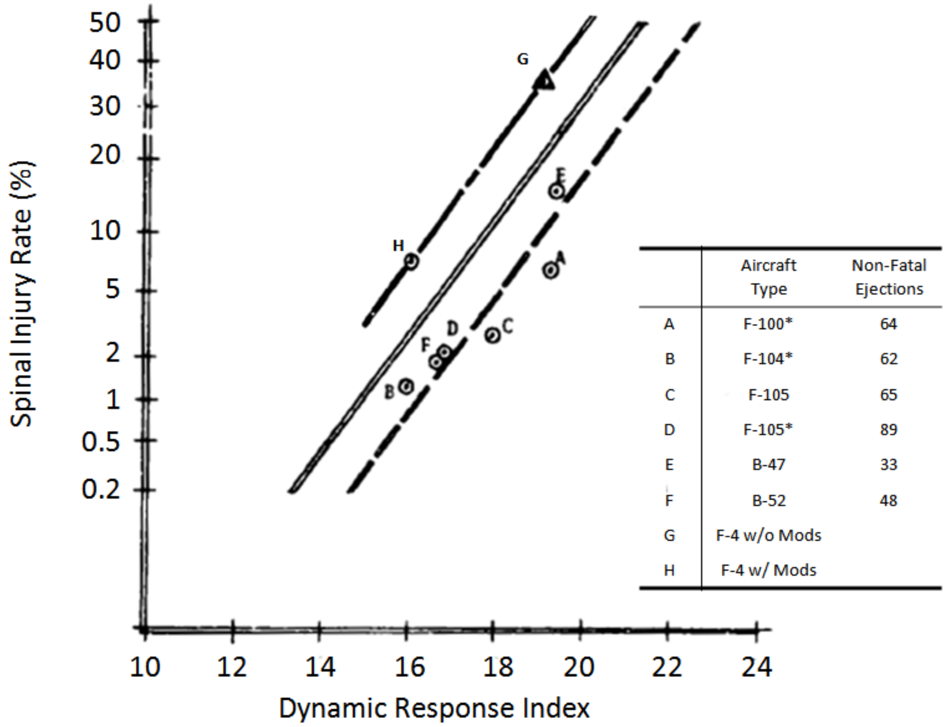


Figure 3-6: Military Aircraft Spinal Injury Risk During Operational Ejections [17]

Collaborative efforts were established in the early years of biomechanical research between the Air Force and NASA due to overlapping efforts. John P. Stapp and his contemporaries established amplitude, onset acceleration and duration of acceleration that resulted in moderate to severe injury. Testing included military volunteers, animal surrogates and data from accidental human exposures [18-20]. Figure 3-7 is an example that summarizes research just for the $+G_z$ limits ranging from non-injured to severe injury [5].

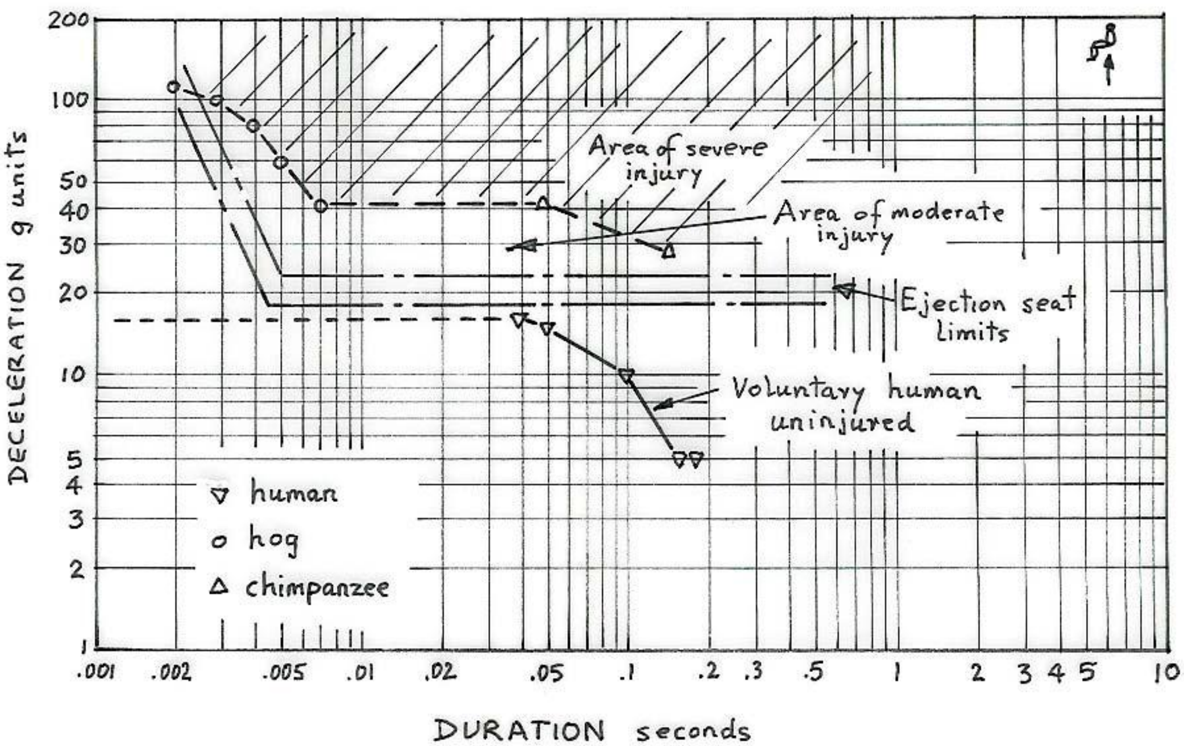


Figure 3-7: Survivable Abrupt Positive G (+Gz) Impact [5]

In 1982, the Air Force Aerospace Medical Research Laboratory (AFAMRL) found injury mitigation in acceleration preloading, defined as an imposed acceleration preceeding the acceleration pulse in the same direction. Table 3-2 outlines, of the 6 baboon cadavers tested on the accelerator (no preload), there was one clavicle fracture, one hepatic laceration and four transections of the rectus abdominis muscles. The decelerator or preload condition (0.25G and 0.962G) however, had one significant injury which could be attributed to preload [21].

Table 3-2: Baboon Cadavers Injuries -50 G_x Impacts [21]

Test Facility	Accelerator	Decelerator
Subjects	6	6
Significantly Injured Subjects	4	1
Fractures	1	1
Muscle Tears	4	0
Liver Tears	1	0

3.1.1.5 Spaceflight Evidence

3.1.1.5.1 US Space Programs

In the US space program, there is very little injury data that is attributable to landing. During the Mercury and Gemini programs, no injuries were reported. During the Apollo program, only one injury was reported, in which a loose item struck a crewmember resulting in a head injury during a 15G landing [22]. However, this injury was not due to the dynamic loading on the crewmember or the interactions with the vehicle, so this injury does not meet the definition of injury defined above.

Since the Space Shuttle was designed to land on a runway with similar dynamic loads to a commercial aircraft, no acute injuries would be expected during the dynamic loads of landing. There is evidence however to suggest that injury can present well after landing: There is a 4.3 times greater incidence of herniated nucleus pulposus occurring post-landing than in control populations, which may be caused by a variety of effects including landing impact [23].

The unfortunate accident of the Columbia mission revealed ineffective occupant protective measures. The *Columbia* Crew Survival Investigation Report cited inadequate upper body restraint and protection as a potential lethal event and recommended that future spacecraft suits and seat restraints should use state-of-the-art technology in an integrated solution to minimize crew injury and maximize crew survival in off-nominal acceleration environments (L2-4/L3-4) and should incorporate conformal helmets and neck restraint designs similar to those used in professional auto racing (L2-7) [1].

3.1.1.5.2 USSR Space Programs

In preparation for the Soyuz program, the former Union of Soviet Socialist Republics (USSR) conducted 130 human volunteer tests. During this testing landing orientations and impact velocities were varied to understand the effect on injury risk (Figure 3-8).

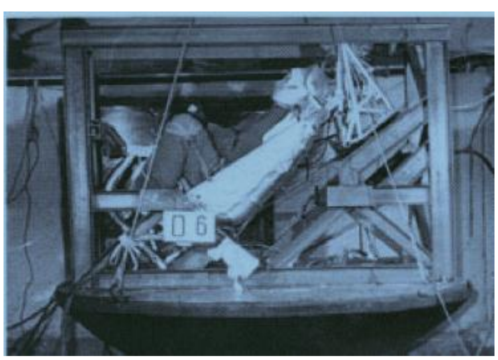


Figure 3-8: Soyuz Drop Test Platform [24]

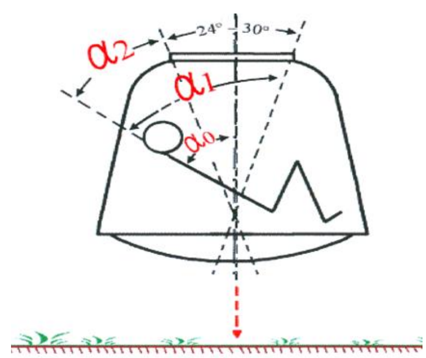


Figure 3-9: Seat Testing at Various Angles [24]

All experiments with high velocity (5-6m/s) resulted in pain in the head, abdomen or hips. In some cases blood was present in urine due to impact. Study cases included testing at angles from 36° to 82° relative to the perpendicular vector from the ground (Figure 3-9). The thoracic spine was at greater risk of injury at 36° (or G_z) while head and organs were found to be at greater risk at 82° (G_x). One 36° tilt test, at 4.5 m/s with no shock absorber, resulted in spinal compression fractures of T4 / T5 Figure 3-10. Investigators later found the subject had scoliosis, which was over looked in the physical.

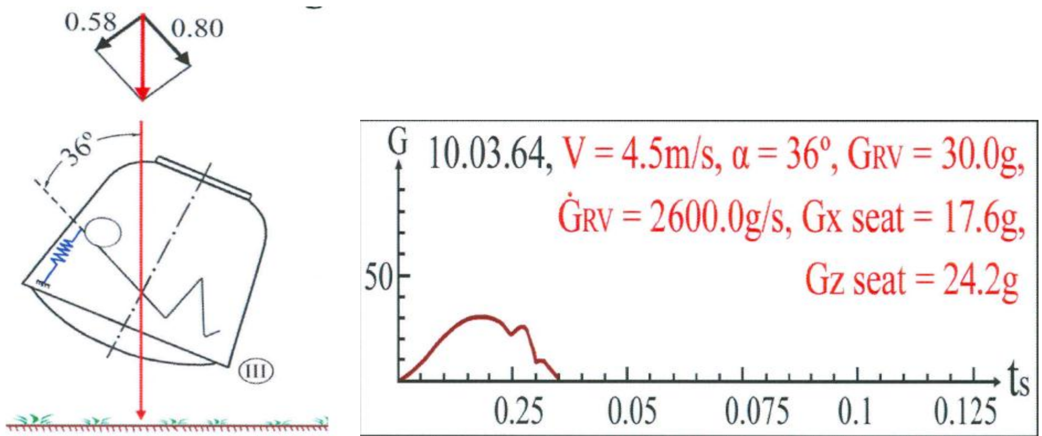


Figure 3-10: Soyuz Test Condition Causing Spinal Injury [24]

Operational experiments from the Soyuz vehicle provide compelling evidence to the risk of injury due to vehicle dynamics for capsule-like landing vehicles. The Soyuz has been operational since 1967 and has completed 112 flights as of August 1st, 2012 with 283 crewmembers. During its 55 year history, there have been 4 fatalities, 1 permanent disablement, 1 moderate injury, and 13 minor injuries reported (see Table 3-3). Three of the fatalities were not as a result of inertial accelerations [25-27]. The number of minor and moderate injuries is suspected to be underreported.

Table 3-3: Number of Injuries During Soyuz Abort and Landing

Soyuz Type	Number of Flights	Number of Crew	Minor Injuries	Moderate Injuries	Severe Injuries	Life Threatening or Fatal Injuries
7K-OK	10	22	1	0	0	1*
7K-TM	29	56	3	0	0	1
T	15	38	4	0	0	0
TM	33	90	3	0	0	0
TMA	22	65	2	1	0	0
TMA-M	3	9	0	0	0	0
Total	112	283	13	1	0	2*

* Does not include Soyuz 11 Loss of Crew (3) because fatalities were not due to landing impact

The dynamics of the landing are variable for each landing, and there have been several landings that were reported to be hard. Table 3-4 shows the relationship between the injury rates during nominal, off-nominal and hard landings [25-27]. As expected, injuries are more likely during hard landings than nominal landings. Although the injuries have been documented, the landing dynamics that caused the injury are not currently available and may no longer exist.

Table 3-4: Incidence of Injury Related to Hard Landings from Soyuz

Injury Rates	Nominal	Off-nominal	Hard	Total
Minor	1%	16%	21%	5%
Moderate	0%	0%	3%	0%
Severe	0%	0%	0%	0%
Life Threatening / Fatal	0%	0%	7%	1%
Number of Crew	218	32	29	280
Occurrence of Crew Landing	78%	11%	10%	100%
Number of Landings	86	13	12	111
Occurrence of Landing	77%	12%	11%	100%

3.1.2 Seat & Harness System

The ability of a vehicle to transport an occupant from one distance to another in a timely manner is of great value. During travel the occupant is moving at the same velocity as the vehicle. However, when an impact occurs, the vehicle quickly decelerates while the body continues to move in the same direction and velocity. The body movement within the vehicle creates an opportunity for blunt contact. To prevent blunt contact, restraint systems are used to couple the occupant to the seat which is coupled to the vehicle. Figure 3-11 provides an example of how automotive restraints minimize displacement and acceleration of occupant during impact versus unrestrained as shown in Figure 3-12 [28]. These figures summarize that the restraint system tied to the seat are critical for occupant protection.

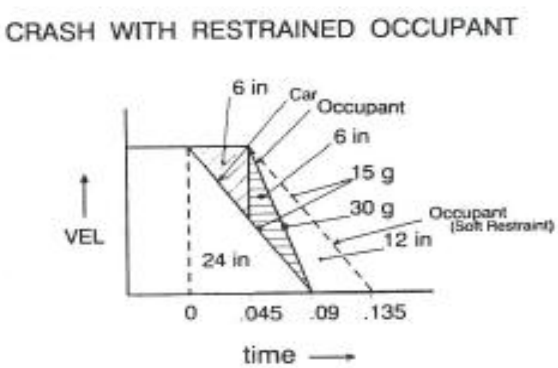


Figure 3-11: Car Crash with Restrained Driver [28]

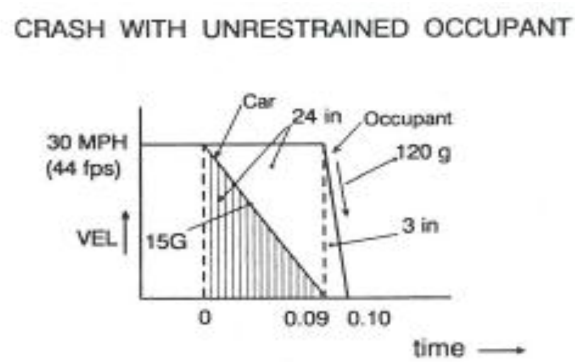


Figure 3-12: Car Crash with Unrestrained Driver [28]

The following studies provide evidence that without optimizing the restraint system the dynamic loading transmitted to the occupant results in injury. Eiband showed that an inadequate restraint/seat system magnified other acceleration parameters (onset, magnitude duration) of the human response [5]. Eiband stated inadequate restraint systems would compromise basic human function that would inhibit emergency egress from vehicle [5].

In 1968, Synder, *et al.* studied types and severity of injuries based on different restraint configurations as shown in Figure 3-13 [29]. The Federal Aviation Administration (FAA) conducted 60 Savannah baboon experiments using five different restraint systems. While the simple lap belt tests were not all fatal, the single diagonal belt tests were all fatal. The lap belt alone has been found to produce congestion and/or minimal hemorrhages in several organs such as the brain, spleen, heart, uterus and pancreas. More severe injuries include ruptured bladders, pulmonary lacerations and interstitial pericapsular renal hemorrhages. The 3-point or double shoulder harness offered more protection than single belts with the disadvantage being that the occupant could slip out of the belt during side impact. Furthermore, there is no protection provided for the cervical spine, risking injury of the neck. Some injuries occurred at high G with scapula fracture and partial dislocation of humerus with the Y-yoke restraint. The conclusion states that the Y-yoke (with inertia reel) and especially with the airbag provided the most protection relative to the other restraint options [30].

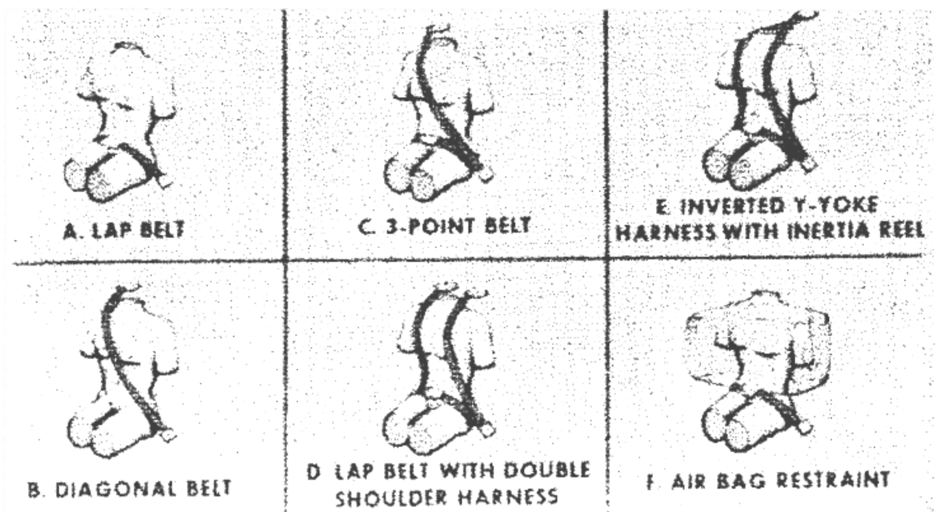


Figure 3-13: Harness Configurations [30]

Zaborowski also conducted deceleration experiments with a restraint combination system during lateral impact with 52 human volunteers [31]. Results illustrated that the restraint system allowed a minimum deflection of 5° during testing. While no permanent damage was observed, minor complaints of sore neck muscles were reported for more than 60% of the exposures above 8 G. An incident did occur in which a subject fainted in the seat, and the blood pressure could not be detected by the sphygmomanometer. Medical monitoring measured 20-30 beats/minute for heart rate, and the subject recovered within 5 minutes once posed in supine position [31].

No matter what restraint system is used, it must fit appropriately to the body to ensure that there is minimal slack in the system to reduce risk of injury [32]. Without proper fit of the restraint system, the human response relative to the vehicle will be amplified. The results from the Space Shuttle *Columbia* accident investigation strongly stated that lethal effects to the crew occurred due to the lack of proper restraint since the inertial reel did not engage during the off-nominal loading. This resulted in inadequate upper body support and allowed the body to swing with only the lower body restrained, resulting in trauma[1].

To optimize restraint systems for protection the performance should include restraining the occupant during onset of impact, distributing loads over anatomical regions of the body that have minimum deflection and minimize body movement relative to the seat [28]. Seat design is an important aspect of occupant protection. Early impact studies (Brinkley) identified the benefits of individually contoured body support that could be formed to fit the occupant by evacuating air from the liner, which was filled with small plastic spheres (Figure 3-14) [33]. The contoured seat behaved as additional restraint system in combination with the harness (Figure 3-15).

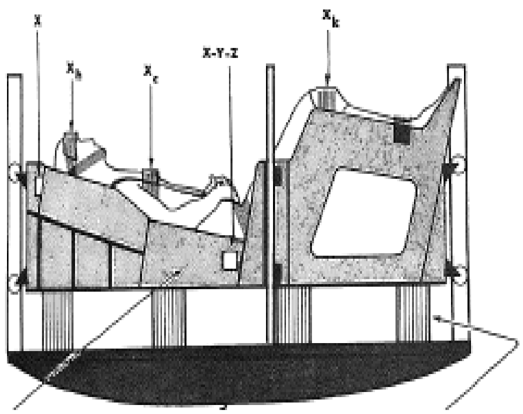


Figure 3-14: Impact Vehicle Test Apparatus [33]

Figure 3-15: Contour Body Support Seat[33]

A similar contour methodology was used by former USSR engineers in seat design to provide "uniform pressure distribution for the human body [14]." From 1963-1967 the USSR performed over 130 drop tests with human volunteers at varying angles of impact and velocity using a shock absorbing actuator (Figure 3-16). Testing concluded this design feature of the seat was critical for supporting the occupant and a conformed liner for every occupant aboard the Soyuz is required to this day.

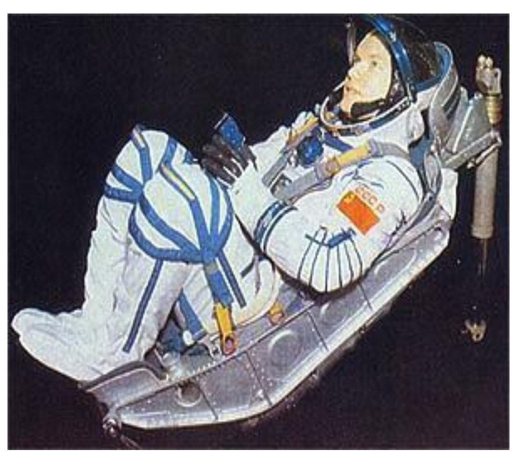


Figure 3-16: Kazbek seat for Soyuz Vehicle [14]

Further human testing was conducted to characterize the dependency of the shock absorbers on the subject's body weight, the effects of the headrest recline angle, the mitigation of suit/helmet effects using a conformal seat, and the efficacy of limb restraints to prevent flail. Modifications were aimed to improve reduction of onset rate and decrease spinal column flexion and neck flexion [14].

The Air Force Research Laboratory (AFRL) has conducted numerous experiments to evaluate the influence of factors such as seat geometry, restraint system design features (including attachment position), and seat cushion properties on the likelihood of injury using vertical and horizontal impact facilities. The cushion properties for example may amplify the impact response by storing up the energy in the seat releasing elastic recoil during impact. Caldwell *et al.* found significant differences ($p<0.1$) in chest displacement across human volunteer testing when comparing the Vertical Impact Protection seat to the ACES II F-16 seat [34]. These effects have been demonstrated by computational models and empirical human testing to be a risk for spinal injury [35]. Further studies include: a study of the effects of seat back angle on impact response [36], and a study of the influence of a negative-G strap to mitigate risk of injury [37].

3.1.3 Spacesuit & Helmet

One of the unique aspects of the NASA environment is the use of a pressurized suit, or spacesuit. This suit is designed to protect the crew from the vacuum of space by providing a pressurized environment around the body, breathable atmosphere, thermal protection, and micrometeorite protection. In addition to these basic functions, there are other considerations in suit design including mobility, suit fit on a wide range of crewmembers, and contingency Extravehicular Activity (EVA). Because the suit must provide all of these functions, it may not be optimized from an occupant protection standpoint. The following studies provide evidence the design of the suit/helmet could induce injury during dynamic loading [38-46].

There are several considerations for the occupant during abort and landings which relate to the suit design. First, the suit, unlike most clothing, may contain rigid elements, which depending on their placement could induce point-loads or blunt trauma resulting in crew injury. For instance, post-mortem human subject (PMHS) studies conducted by NASA at Ohio State University (OSU) investigated the effect of rigid suit elements during landing impacts [38]. Although an insufficient number of PMHSs were tested, the results clearly indicated that the rate of injury from poorly placed suit elements, such as ring placement, drastically increases the risk of injury [39].

Another rigid body on the suit is the Suit Mounted Connector (SMC) which includes supply and return lines for breathing air, cooling water, suit power, and communications for the crew member. Wake Forest University conducted impact simulations of a model to investigate the human response to the mass, shape and placement of the SMC [40]. The design locations were evaluated using the test matrix show in Table 3-5. Figure 3-17 shows the two proposed mounting locations, the chest and the thigh.

Table 3-5 Chest Injury Risk

Simulation	V*C (m/s)	Deflection (%)
No Umbilical	0.3635	-20.71%
5 lb, 1 mm	0.7476	-29.87%
5 lb, 15 mm	1.0785	-30.43%
5 lb, 30 mm	1.2240	-30.30%
7 lb, 1 mm	0.8575	-34.78%
7 lb, 15 mm	1.0126	-35.72%
7 lb, 30 mm	1.2614	-36.14%

If V*C > 1, >25% risk of AIS 4+ injury

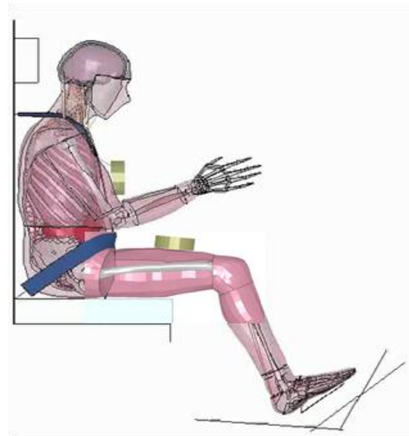


Figure 3-17: SMC Mounting Locations

The analysis found that the thigh location had a negligible effect on the risk of injury; however, the chest mounted connector simulations showed the potential for severe injury, as summarized in Table 3-5 [40, 41]. Minimizing chest compression not only reduces risk of fracture, but reduces the risk of commotio cordis, which is a circulatory arrest due to a non-penetrating impact to the chest and which could result in sudden death [42]. Therefore, the placement and design of suit components is critical to protecting the crew during dynamic loading.

Another rigid element of the suit that poses a risk is the non-conformal helmets, which is unique to the spaceflight industry. While other industries (automotive racing, military, sports) design their helmets to mitigate energy transferred to the neck and spine, the spaceflight helmet incorporates more objectives. As a result, design of spaceflight helmets may not be optimized for occupant protection.

Radford et al. conducted a study looking at the suit as a whole during +Z accelerations and with a Hybrid III FE model (see Figure 3-18) [43]. From this analysis, the head mounted mass was found to be a concern, as the helmet approximately doubled the neck compression force compared to the unsuited case as shown in Table 3-6.

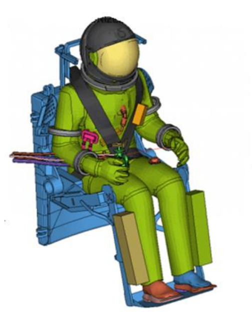


Figure 3-18: Suited Model

[43]

Table 3-6: Suit Effects on Neck Compression

Load Condition	Probability of Occurance	Peak Neck Compression Unsuited [N]	Peak Neck Compression Suited [N]
Nominal	92.9%	690	1,400
Nominal	92.9%	500	1,300
Off-Nominal	6.7%	800	1,900
Off-Nominal	6.7%	1,200	2,100
Off-Nominal	0.2%	960	2,200

Yoganandan *et al.* report injuries from neck compression with as little as 1,100 N of compressive force [44]. If the spine is not aligned the risk of injury increases considerably [45, 46]. This was determined operationally on the F-4 ejection seat, where the spine was misaligned and resulted in a 34% rate of injury versus predicted injury by Brinkley Model if the spine was aligned at 5% risk [17, 27, 47]. ILC Dover, NASA, Gentex Corporation and Hamilton Sundstrand Helmet researched design considerations for the helmet that maintained visibility inside and outside the vehicle as well as a helmet designed for protection during landing. Recommendations were to reduce the mass of the helmet, secure the helmet to eliminating the neck from holding the load, and provide a foam collar for neck. Another possible design was a conformal helmet. [48]. These recommendations are consistent with the *Columbia* Crew Survival Investigation Report which cited several potentially lethal events and recommended countermeasures to improve the survivability in the future. One of the five potentially lethal events identified was the nonconformal Advanced Crew Escape System (ACES) helmets do not provide adequate head protection or neck restraint during dynamic loading. Recommendation L2-7 from the report states: "Design suit helmets with head protection as a functional requirement, not just as a portion of the pressure garment. Suits should incorporate conformal helmets with head and neck restraint devices, similar to helmet and head restraint techniques used in professional automobile racing [1]."

An additional challenge of occupant protection is restraining the body in the case of a pressurized suit. In the case of landing with the suit inflated, additional movement of the body inside of the suit may occur during impact. In this case, the vehicle restraint system is no longer restraining the crewmember, but is instead restraining the suit allowing the occupant to move freely inside the suit [28]. Kornhauser had one case of a fracture in the seventh thoracic vertebra which occurred as a result of impact testing with the pressurized suit partly inflated [49]. This could be analogous to a loose restraint system. Other investigators found through experimentation that severe and persistent pain was experienced by subjects as a result of a loose restraint system increasing risk of injury [18, 30].

3.2 Intrinsic Injury Risk Factors

Currently, NASA vehicles must be capable of accommodating 1st percentile females to 99th percentile males [50]. No limit exists for age. Protection for such a wide population is a challenge as most occupant protection data is either based on young, male military subjects or elderly male post-mortem human subjects (PMHS). Therefore, threshold limits specific for the astronaut corps remains an open issue since there are biomechanical differences related to gender, anthropometrics and age.

3.2.1 Age

As the human body ages, tissue properties change, e.g., the yield point, Young's Modulus and human tolerance during dynamic loading are lowered. Evidence that age changes tissue properties is well documented. Figure 3-19 illustrates bone strength begins to degrade after the age of 39 years old. Other anatomy such as the intervetebral disc degenerates after the age of 25 [28]. Pintar *et. al.* found that the Young's Modulus of the anterior cruciate ligament of young specimens (16-26 years) was markedly higher at 111+/- 26MPa as compared to older specimens (48-86 years) with a value of 65+/- 24MPa [51]. Muscle is another tissue that changes with respect to age. Foust *et. al.*, studied cervical spine of 180 volunteers ranging in age from 18-74 years old. In comparison older volunteers had a reduced range of motion as much as 40% , muscle reflex reduction of 23% and strength loss of 25% [52]. The same concept applies to other sections of the body such as thoracic, abdominal, pelvis, cervical and extremities [28, 53, 54]. Figure 3-20 illustrates the probability of cervical spine failure with respect to age the spine at a loading rate of 2.2m/s.

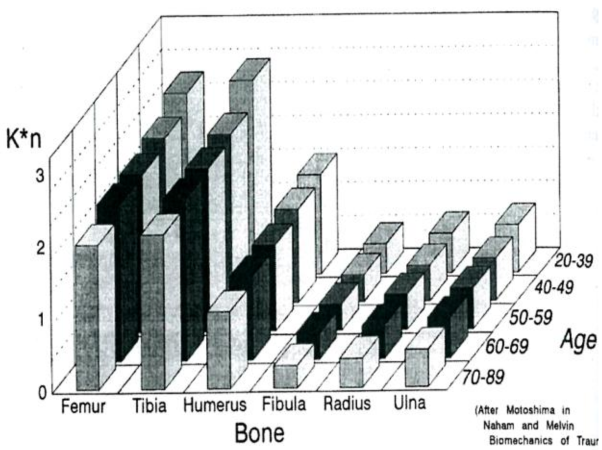


Figure 3-19: Bone Strength Decreases with Age [28]

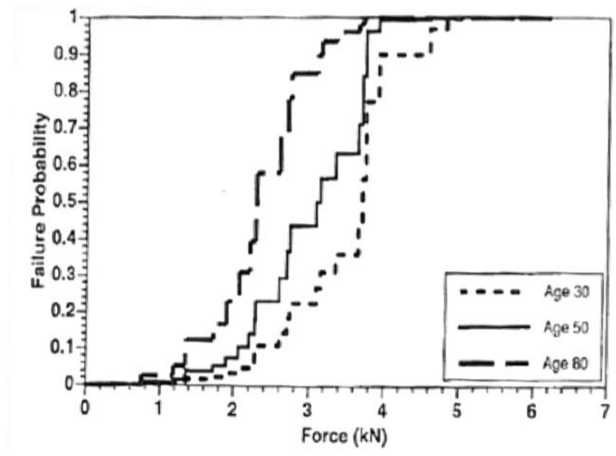


Figure 3-20: Failure of Male Cervical Spine. [55]

Further evidence exist that the risk of injury due to dynamic loading increases with age. Fatality Analysis Reporting System (FARS) analyzed accidents from 1975-1998 from the National Highway Traffic Safety Administration (NHTSA). Conclusions state the increase risk of death due to age for the same blunt trauma increases 2.52% for males and 2.16% for females per year after the age of 20 [56]. Little information is available in automotive industry, and even less is known in spaceflight industry. Therefore, gaps of knowledge remain in Occupant Protection of what risks are attributed to age during dynamic loading with spaceflight profiles.

3.2.2 Gender

The risk of injury during dynamic loading has been shown to be significant between gender differences. Epidemiology studies for the automotive industry use accidental crash information to better understand methods of improving countermeasures for the driver. Given the same automobile accident, the fatalities for women were 22%-25% greater than for males [57, 58]. The risk of injury in the thoracic spine has been found to be 20% greater in females then males.

Allnutt conducted a review of medical and safety literature and reported females to have differences in density, structure, size and strength of bone when compared against a well-matched control cohort of males of the same age, height and weight. This is in part due to the female's bone structure, which has a thinner cortical layer relative to the trabecular section of the bone as compared against males [59].

Another study compared tomography scans of cervical spine at C4 from matched sized volunteers. Significant differences were determined through analysis of variance [60]. Gallagher *et. al.* quantified 14-18% greater stress in the cervical spine during dynamic load of 10 when compare to males [61]. Gender differences require further investigation to fully understand the risk of injury due to dynamic loading in order to better protect vehicle occupants.

3.2.3 Anthropometry

Anthropometry of a person is highly critical when it comes to fitting the occupant to the seat with the restraint system. If the restraint/seat/suit/helmet configuration is not optimized for the occupant, the risk if injury increases [32]. Anthropometric measures that include the length of the spine present challenge for spaceflight, since this is altered due to gravitational changes and fluid shift [62]. Spaceflight has found 4-6cm increase in body height measure in crew [63-65]. Bed rest studies found lumbar spinal length to increase up to 3.7+/-0.5mm with a decrease in spinal curvature [62, 66]. Figure 3-21 compares angle between the L1 and S1 as well as the disc height in L2/3 viewed in the sagittal plane pre and post bed rest. Conclusions state lengthening of the spine, increase disc size and flattening of spinal curvature [62]. Research is ongoing to characterize spinal changes during spaceflight which will be critical for occupant protection countermeasures.

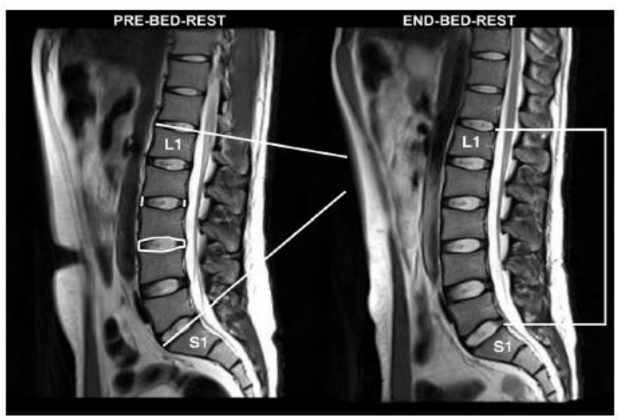


Figure 3-21: Bed rest MRI of the Spine Before and After Bed rest. Note the change in angle between the L1- S1 and the disc height in L2/3 viewed in the sagittal plane [66].

3.2.4 Spaceflight Deconditioning

During spaceflight, musculoskeletal systems change in structure and function due to unloading of the body in microgravity environment over time. During prolonged spaceflight, skeletal density changes are seen, primarily in the lower extremities and spinal elements [67]. Studies conducted using dual energy X-ray absorptiometry (DXA) have shown decreases on average of 1-1.6% in the spine, femoral neck, trochanter, and pelvis, with an average of 1.7% in the tibia after only one month in microgravity [68, 69]. Because skeletal deconditioning is time dependent, any method for accommodating the losses will be mission length specific.

In addition, changes in muscle mass and strength occur, and are dependent on the exercise regime employed during spaceflight. During Skylab missions, leg volume decreased by 7-10% [70] as well as up to 19% in crewmembers aboard the MIR space station [71, 72]. The muscle loss experienced by crewmembers is also selective; muscle fiber size in the vastus lateralis (VL) decreased after 5-11 days in flight at different rates. Edgerton *et al.* found decreases of 16% in Type I, 23% in Type IIa, and 36% in Type IIb fibers [73, 74]. Tendon tissue, which attaches the muscle to the bone was also studied using unloading models (Unilateral Lower Limb Suspension and Bed Rest). The results concluded an increase of Young's Modulus in the tendon resulting in muscle shortening, which negatively affects muscle function and performance [75]. Changes in cross-sectional area of intervetebral discs and overall shape of the spine are attributed to microgravity environment [23]. Current studies are in place to further investigate intervertebral discs during spaceflight; however no research currently addresses the risk of injury during dynamic loading for a deconditioned spine.

4.0 INJURY RISK ASSESSMENT METHODS

There are 3 main categories of methods for assessing injury risk due to dynamic loads. The categories are: humans, human surrogates, and numerical models. As seen in Figure 4-1, each category (indicated by color) has several possible method of assessment. Regardless of the method chosen to assess injury, criteria must be defined to relate responses to injury risk.

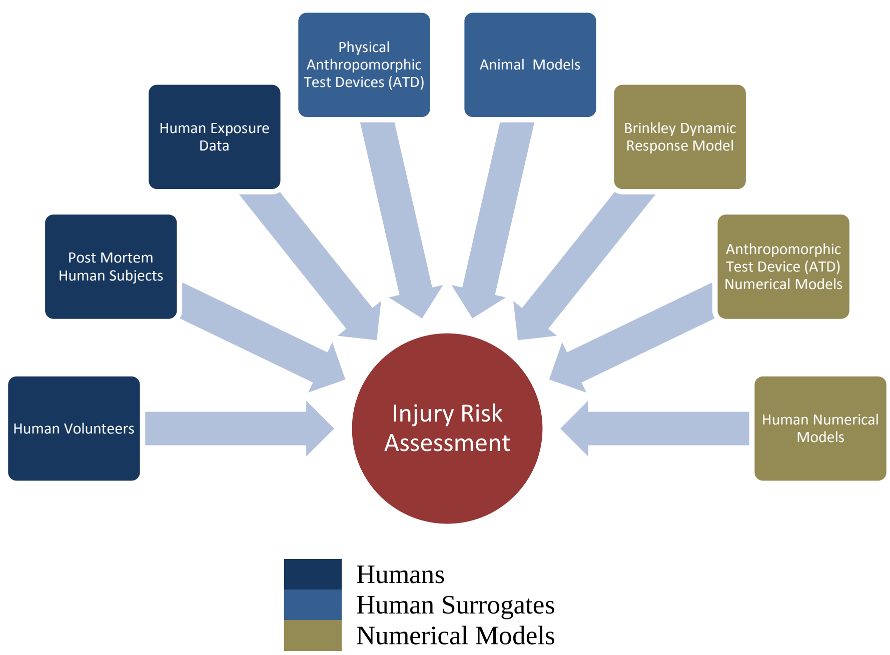


Figure 4-1: Available Injury Assessment Methods

4.1 Humans

4.1.1 Human Volunteers

To understand injury tolerance levels for crew members, the obvious choice is to conduct human volunteer testing post spaceflight mission. However, this would be unreasonable. One option is to conduct testing with healthy human volunteers at non-injurious dynamic loads. The information would be most accurate, but does not come without complications. This data would provide whole body human tolerance curves but it would be unethical purposefully test at outcomes for minor risk injury curves [76]. Human volunteer testing at noninjurious levels also provide challenges due to high cost, limited testing facilities in the United States, expertise, and increase of time required for testing humans.

4.1.2 Post Mortem Human Subjects

Post Mortem Human Subjects (PMHS) or cadavers are another option for assessing injury risk. Since PHMS are humans, their anatomy and anthropometries are human. One of the greatest advantages of PMHS testing is the ability to more accurately pinpoint the threshold at which a human injury would occur. This can be accomplished by imbedding sensors in the body to directly measure forces, accelerations, and moments, as well as with post test autopsy. These data allow direct determination of injury risk in specific anatomical regions. PMHS also serve as a valuable tool to devise Anthropometric Test Devices (ATDs) and computational models [76].

Although there are many advantages with PMHS testing, there are also limitations. Because of limitations in the availability of PHMS, subjects may not be representative of the astronaut population in age and overall fitness. In addition, positioning of PMHS for testing can be difficult as PHMS do not have active muscles to maintain an upright posture in a seat. A lack of active muscle contractions, differences in tissue properties, and differences in tissue responses may affect the measured responses, thus affecting the assessment of injury risk for a live human. Finally, working with PMHS limits the number of facilities for testing and complicates the use of equipment (i.e., suits that cannot be reused after testing) [77].

4.1.3 Human Exposure Data

Human exposure data are data collected where humans are inadvertently subjected to injurious conditions during otherwise routine events in life. Some examples are automotive crash data, automotive racing impacts, and military aircraft mishaps. Since the events that produce such data are undesirable, every effort is made to prevent such situations. Even so, the events still occur and in some instances are well documented.

Human Exposure data may provide information unattained in the laboratory such as intrinsic comparison (age, gender, anthropometrics) and multidirectional dynamic loads [56-58, 78, 79]. However, details of the incidents are critical to evaluate if the data is applicable to spaceflight conditions. For instance, if neck injury were to occur during an emergency evacuation from an aircraft via seat ejection due to a failure of the canopy removal system, the neck would expect loads outside of nominal spaceflight conditions [34]. Therefore, this data would not be applicable to spaceflight scenarios.

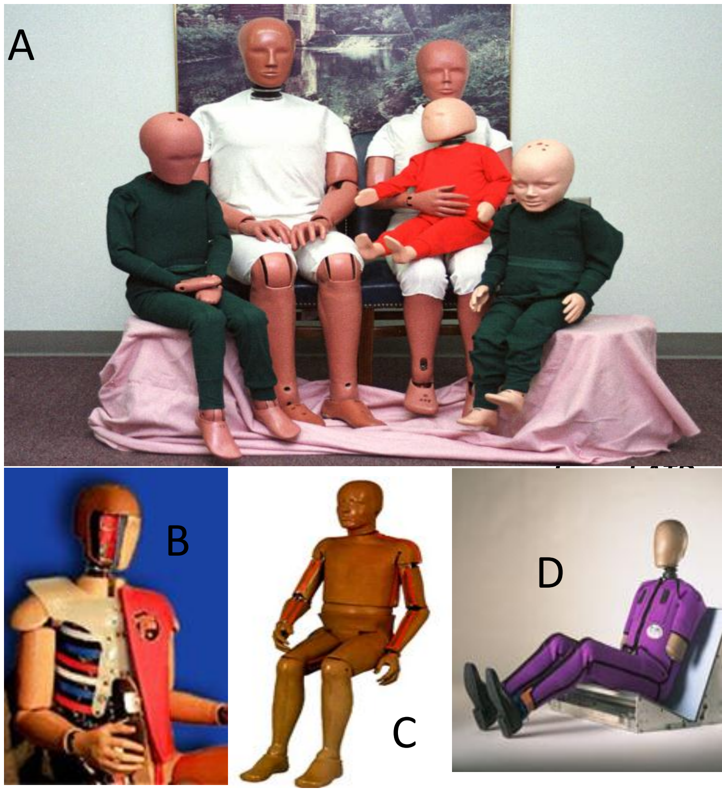
4.2 Human Surrogates

4.2.1 Physical ATDs

Anthropomorphic Test Devices (ATD), also known as crash test dummies or manikins, have been used for decades to assess injury risk to humans in specific impact scenarios. Originally, ATDs were used for military aircraft injury mitigation, and are now commonly used in the development and verification of safety measures for a variety of transportation systems. The purpose of ATDs is to replicate human responses to particular impact situations and offer repeatable responses. This is a significant advantage over previously discussed assessment methods, which are prone to significant inter-individual variability.

Although this is the goal, often other factors prevent the ATD from responding the same as a human. First, ATDs are designed to withstand higher forces than a human so that they may be reused. In addition, many simplifications are necessary in the anatomy of the ATD to allow cost-effective design and construction. Since ATD do not always respond as humans would, injury risk functions are used to relate the ATD responses to actual human injury. This application might not be optimized for minor injury detection or human tolerance since the ATD does not provide discomfort/pain feedback.

ATD span a wide range of purposes, sizes and applications. The automotive industry has a large variety of various ATDs that are available for different directions of impacts and occupant size. Figure *4-2* shows a variety of ATDs developed for different uses and anatomical sizes.



thropomorphic Test Devices (ATD) A) tal Impact Family (L to R: 10 year old, 50th le adult, 5th percentile female adult, 3 ar old), B) THOR 50th percentile Frontal) ADAM 95th percentile Military Vertical nd D) WorldSID 50th percentile Side y (SID).

4.2.2 Animal Models

Animal models have been used extensively in the past and have several advantages and limitations. Clearly, animals offer the unique advantage of studying living tissue response. In some cases a combination of surrogates are required to determine countermeasures. While PHMS data may be used to determine brain motion and deformation, it does not provide live physiological response such as minor traumatic brain injury, which takes time to develop after impact [76]. This information may be further used to develop mathematical models specific to research needs. Since animals are not anthropometrically similar to humans, only trends may be identified relative to human response [80].

4.3 Numerical Models

4.3.1 Brinkley Dynamic Response Model

The Brinkley Dynamic Response (BDR) criteria were developed as a result of an evolutionary process to define the human dynamic response and risk of injury. The BDR is a simple, lumped parameter, single degree of freedom model, which is intended to predict the whole body response to acceleration as shown in Figure 4-3. The body response is calculated with input to acceleration at the seat [17].

Once the dynamic response is calculated, the Brinkley Dynamic Response model is used to calculate β_{max} which predicts approximate injury risk shown in Table 4-1 for each risk level.

Because the BDR model is a simple, lumped-parameter, single degree of freedom model, it only predicts ranges of injury risk for any injury, and cannot provide information as to the severity or anatomical location of an injury.

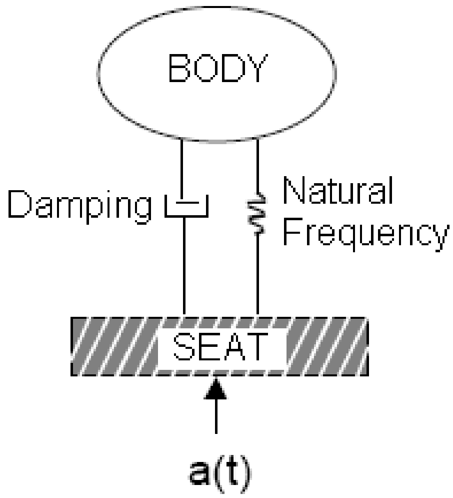


Figure 4-3: Lumped Mass Diagram Of the Brinkley Dynamic Response Model

Table 4-1: Approximate Injury Risk

Risk Level	Approximate Risk
Low	0.5%
Moderate	5%
High	50%

A second limitation stems from the assumption that the spine is in alignment with the acceleration vector Gz. If the spine is 5° out of alignment relative to the load vector, the risk of injury increases dramatically. This was determined operationally on the F-4 ejection seat, where the spine was misaligned and resulted in a 34% rate of injury (5% risk of injury was predicted) [17, 47].

The +Z axis BDR is anchored on operational ejection data based on injuries sustained in the thoracolumbar spine; however, testing for the other axes using the BDR (±X, ±Y, and –Z) were assigned injury levels without statistically based methods [17, 81]. Mr. Brinkley has also expressed concern regarding the Y axis model and warns that the Y axis model for unsupported lateral loads is not correct [82]. Since the BDR model was developed based on simple acceleration profiles, using the BDR model as a stand-alone may not apply because of the complex loads expected for MPCV and other future spacecraft.

Brinkley (1985) expected that different dynamic models would be necessary for changes in the seat and restraint configuration. The BDR model was developed with minimal gaps between the seat supporting surfaces and the test subjects. Additional gaps can allow increased contact forces and increased risk of injury. Because the model treats the whole body as a lumped mass, the seat geometry and restraints used on the test data are critical to achieve the same results. The implications of these limitations are twofold. First, they do not account for improvements in restraint systems, which have been significant over the last 25 years. The consequence is either an overly conservative design, or a design that is not as protective as possible, since no seat design improvements are reflected in the BDR model results. This was shown operationally in Royal Air Force ejection injury rates that were not predicted by the DRI

[79, 83]. In addition, with the seat and restraint system the BDR has no way of accounting for the current spacesuit/helmet donned by the crew. The original BDR model was developed with minimal head supported mass (helmets which weighed less than 5 pounds). Additional helmeted mass (which is probable given NASA's current designs) may cause the natural frequency and damping parameters of the human to change, invalidating the model. In addition, increased head supported mass poses a real risk to the neck due to compressive loading during +Z accelerations, which are not accounted for in the BDR model [43]. Furthermore, rigid elements on the suit must be accounted for in the model to accurately predict injury. Results from suit testing performed by NASA at Ohio State University found that the rate of injury resulting from poorly placed suit elements drastically increases the risk of injury, which the BDR model did not predict [38].

Finally the BDR model also lacks fidelity in regards to variation in gender, weight, anthropometrics and age. The BDR model is representative of human response from young, healthy military personnel which is not only a misrepresentation of the crew population but does not factor microgravity effects or deconditioning status of the crew's health.

4.3.2 ATD Numerical Models

As discussed previously, physical ATDs have several advantages and disadvantages, which are shared with ATD numerical models. In addition to the physical, numerical models offer the ability to test various configurations, loads and responses that are not easily tested with the physical ATD. Thus, numerical models of ATDs offer the advantage of simulating complex testing and assessing hardware without the need to fabricate prototypes. However, ATD numerical models are sensitive to initial conditions. Sensitivity studies are needed to understand how sensitive the responses are to variations in these initial conditions. Some initial conditions that may be important are: initial position of the ATD in the seat, initial tension in the restraints, friction coefficients between the seat and ATD, pre-deformation of the ATD into the seat, and gaps between the ATD body regions and seating surfaces.

Several popular numerical solvers are currently available. The majority of solvers are Finite Element (FE) solvers. Popular choices include LS-DYNA®, RADIOSS®, and PAM-CRASH®. Each solver has different behavior but with some work, FE models can be ported between environments. Within these environments, FE models of various ATD are available with varying degrees of fidelity and performance. MAthematical DYnamic MOdel (MADYMO®) is another solver which uses ellipsoid representations of physical structures to estimate responses. In addition, MADYMO offers the ability to interface with FE models which allows co-simulation with more complex structures. Within MADYMO are a range of models for many different ATD models. Several popular ATD numerical models are shown in Figure 4-4.

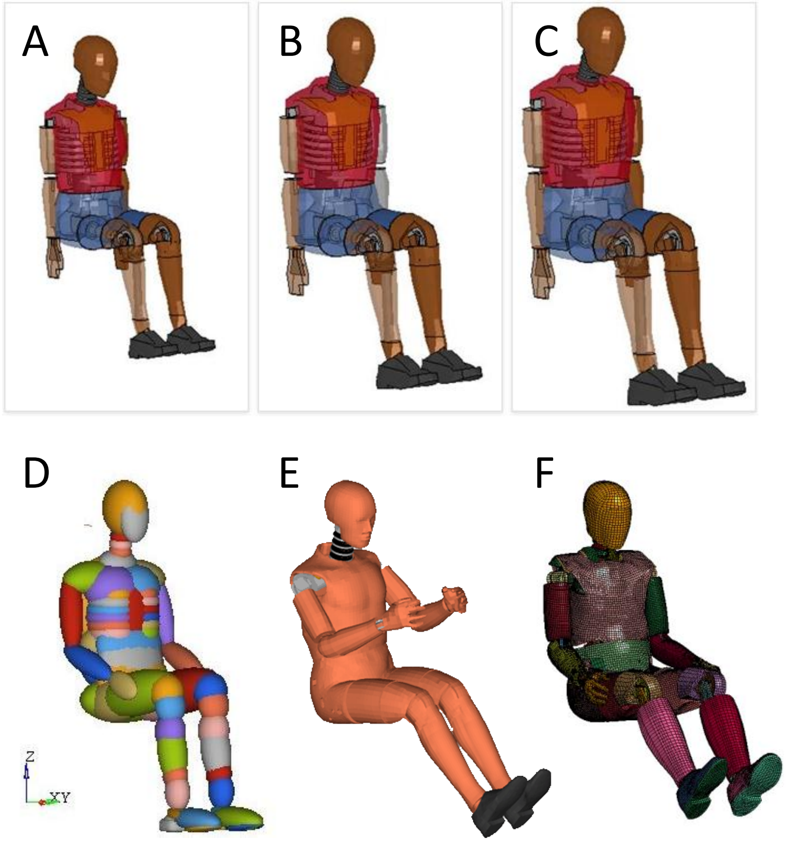


Figure 4-4: Various Anthropomorphic Test Device (ATD) Models. Shown are A) Livermore Software Technology Corporation (LSTC) Hybrid III 5th percentile female LS-DYNA FE model, B) LSTC Hybrid III 50th percentile male LS-DYNA FE model, C) LSTC Hybrid III 95th percentile male LS-DYNA FE model, D) MADYMO Hybrid III 50th percentile male ellipsoid model, E) Humanetics Hybrid III 50th percentile male LS-DYNA FE model, and F) National Highway Traffic Safety Administration (NHTSA) THOR-NT 50th percentile male LS-DYNA FE model.

4.3.3 Human Numerical Models

4.3.3.1 Available models

4.3.3.1.1 Total Human Model for Safety (THUMS®)

THUMS® is a group of Finite Element (FE) models developed by Toyota, as shown in Figure 4-5, which represent a total human including a biofidelic skeleton, muscle and ligament tissues, and internal organs. Currently there are several models of interest including an American mid-sized (50^{th} percentile) male, an American small (5^{th} percentile) female, and an American large (95^{th} percentile) male.

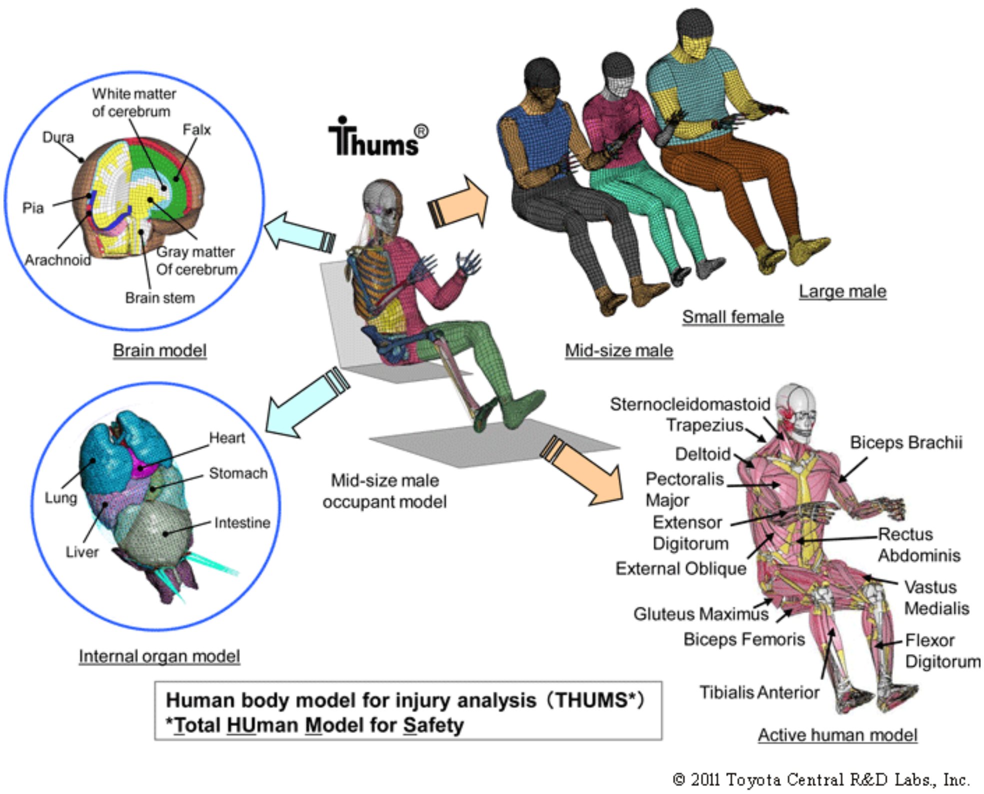


Figure 4-5: THUMS Model [84]

4.3.3.1.2 Global Human Body Model Consortium (GHBMC)

The Global Human Body Model Consortium (GHBMC) is a consortium of auto makers, suppliers, universities, and governments with the goal of creating a single human body model for advancing crash research technology. In 2011, the GHBMC released a 50th percentile male model and plans to develop a 5th and 50th percentile female and 95th percentile male model in the future. The models include detailed anatomical features as shown in Figure 4-6.

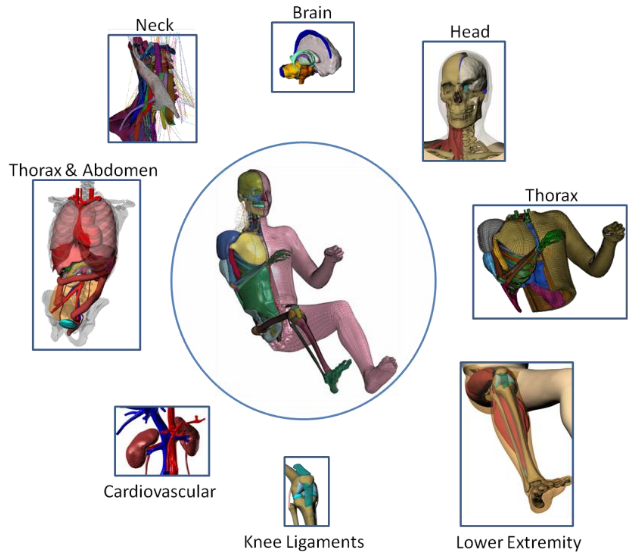


Figure 4-6: GHBMC Human Model

4.3.3.1.3 ESI Human Model

ESI has a line of four human models. Each is a representation of a 50th percentile American male, and represents varying levels of fidelity (Figure 4-7). The AM50a is an articulated rigid body model with rigid body segments and articulated joints. The AM50s has deformable ribs, simplified organs and flesh and comes in both sitting and standing postures. The AM50d is still under development and will be a full deformable human model with modular segments.

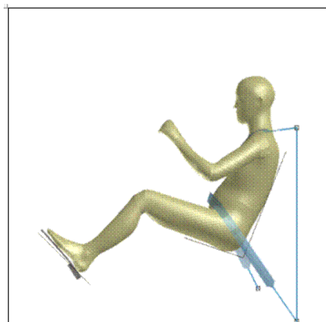
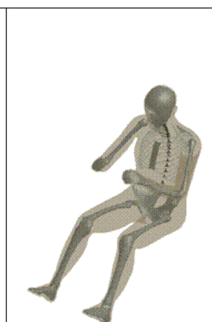
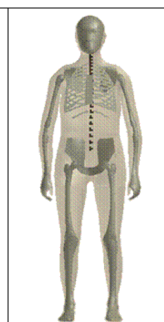
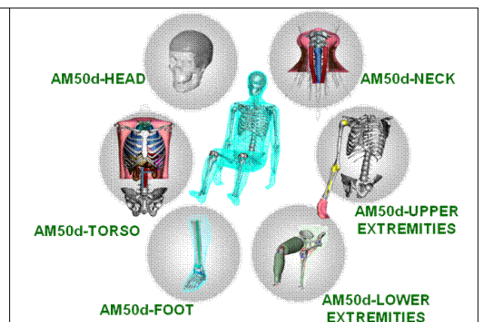
			
AM50a	AM50s - Occupant	AM50s - Pedestrian	AM50d - Occupant

Figure 4-7: ESI Human Models

4.3.3.1.4 MADYMO Human Model

MADYMO human models are available with active muscle control (in the 50th percentile male model) and with passive musculature (5th percentile female, 50th percentile male, and 95th percentile male).

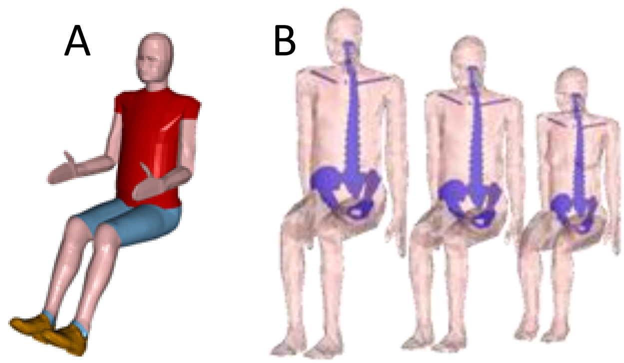


Figure 4-8: MADYMO Human Models.A) Active muscle control 50th percentile human model B) Passive muscle control 5th, 50th and 95th percentile human models.

4.3.3.2 Advantages & Limitations

Human models are a developing field of research and offer great potential to address many of the limitations found in the other methods. Human models can be developed to simulate a variety of intrinsic factors identified previously. They can be developed to account for anthropometry, gender, age (through material property modifications), and possibly even spaceflight deconditioning in the future. In addition, human models can represent soft tissue, internal organs, and the skeletal system, allowing detailed investigations of injury potential to these areas. Because they are anatomically and anthropometrically correct, they can be positioned just as a human could in a restraint system. Finally, unlike human volunteers, human models can be subjected to injurious conditions without harm, and can even simulate tissue failures (e.g., bone fractures).

Although human models may one day eliminate the need for other methods of assessment, currently, the technical readiness level is low. Even with the human models available today, they are being developed for automotive impact cases and aren't necessarily validated in other orientations. In addition, human volunteer, PMHS, and animal data are needed to inform the models to allow accurate simulation. For more detailed injury prediction, much more data is needed.

4.4 Injury Criteria Definition

Regardless of which method is chosen, injury criteria are needed. These criteria can be tolerance limits, defined by non-injurious testing results, or these criteria can be Injury Assessment Reference Values (IARVs) which relate a particular response to injury risk. Either way, these tolerance limits or IARVs must predict injury in a range that is appropriate for the application. NASA currently defines injury risks to be

<0.5% for nominal landings and 5% for off-nominal (based on the Brinkley Dynamic Response Model). Even at 5% risk of injury, most current injury risk functions for ATDs or numerical models are not validated. Most are validated for serious injury and (AIS≥3 or AIS≥4) with a higher risk of injury (15-50% risk) [85].

4.5 Summary

Table 4-2 summarizes options to evaluate risk of injury that include human testing, human surrogate testing or numerical model simulations. Since each method has distinct limitations, no one model can address all of the injury risk factors.

Human testing provides quantitative values in parallel with perception of tolerance for human volunteers and actual human exposure, but testing can only be conducted at sub-injurious levels. Post Mortem Human Subjects (PMHS) do not provide perception of tolerance but can provide direct measures of tissue responses during dynamic loading. If human testing is not required, human surrogates and numerical models can provide valuable information concerning risk of injury due to dynamic loading.

Human surrogates are used to predict injury risk based on correlated responses with humans. Anthropomorphic Test Devices (ATD) for instance can provide mechanical measures during different loading conditions, but they lack physiological and biofidelic responses of a human. One limitation is the lack of local injury, such as point loads or blunt trauma during impact. Animal models provide valuable physiological trends in different testing configurations but obviously require results to be scaled to represent human response.

Numerical models are derived from human and human surrogate data. Therefore, the models are only as accurate as the data that was used to develop the model. Models vary in their level of fidelity (anatomy, physiologic response, direct observation of injury) and technology readiness level (TRL). Some models that are better validated have a high technology readiness level, while models with lower technology readiness levels are not as well validated and may not accurately represent human responses in all conditions. In addition, models with higher fidelity can address more injury risk factors, compared to those with lower fidelity. These models continue to enhance their development of high fidelity transfer functions utilizing technological advances in computational simulation software and testing instrumentation. One or a combination of models may be required to assess injury risk to crewmembers.

Finally, injury criteria must be validated for the desired level of injury risk and severity. Only IARVs or tolerance limits validated to the injury risk level defined are useful for assessing injury risk.

Table 4-2: Relative Strengths and Weaknesses of Each Injury Assessment Method [77] [1]

	Humans			Human Surrogates		Numerical Models		
	Human Volunteers	PMHS	Human Exposure	ATD[2]	Animal	Brinkley Dynamic Response Model	ATD[2] Numerical Model	Human Numerical Models
Extrinsic Injury Risk Factors								
Vehicle Dynamic Profile	Yes	Yes	No[3]	Yes	Yes	Yes	Yes	Yes
Seat & Restraints	Yes	Yes	No[3]	Yes	No	No[4]	Yes	Yes
Suit & Helmet	Yes	Yes	No[3]	No[5]	Partial	No[4]	Yes	Yes
Intrinsic Injury Risk Factors								
Age	Yes	No[6]	No[3]	No	No	No	No	Possible[7]
Gender	Yes	Yes	No[3]	No	No	No	No	Yes
Anthropometry	Yes	Yes	No[3]	Yes	No	No	Yes	Yes
Spaceflight Deconditioning	No	Possible[8]	No	No	Yes	No	No	Possible[7]
Other Considerations								
Anatomy	Yes	Yes	Yes	Partial	No	No	Partial	Yes
Physiologic Response	Yes	No	Yes	No	Yes	No	No	Yes
Injurious Testing	No[3]	Yes	Yes[3]	Yes	Yes	Yes	Yes	Yes
Direct Observation of Injury	No	Yes	Yes	No	Yes	No	No	No
Technology Readiness Level[9]	High	High	High	High	High	High	Moderate[9]	Low

[1]Adapted from Crandall, et al. [77]
[2]Anthropomorphic Test Devices
[3]Not possible prospectively
[4]The Brinkley Dynamic Response Model was validated using specific seat and restraint setups and dynamics. The model may not predict injury accurately when extrapolating beyond this setup and dynamics.
[5]Not possible to assess localized injury potential
[6]Although possible prospectively, very difficult in practice due to limited subject pools
[7]Currently Available Human numerical models do not specifically address these factors, but could be modified to simulate the increased risk of injury
[8]Selection criteria could be used to select only subjects with similar bone mineral density (BMD), although this is not a true representation of spaceflight deconditioning.
[9]Technology Readiness Level (TRL) is a measure of how ready each method is for immediate use. ATD models are at various levels of TRL depending on the solver, ATD family and size

5.0 RISK IN CONTEXT OF EXPLORATION MISSION AND OPERATIONS

Given the intrinsic factors identified, only one is affected by exploration missions and operations. Spaceflight deconditioning has been found to be related to dwell time in reduced gravity environments, thus without appropriate countermeasures, the risk of injury due to dynamic loads could increase. This is assuming that no other extrinsic factors have changed.

The extrinsic factors identified, while not directly affected by mission length or destination, can be used to mitigate the injury risk associated with spaceflight deconditioning.

6.0 GAPS

Based on the evidence presented above, several knowledge gaps have been identified. These can best be discussed based on the related risk factors and assessment methods.

Based on the evidence presented, significant research has been conducted in the X and Z axes; however, very little has been conducted in the anticipated orientations and complex dynamics expected in spaceflight. Additional research may be warranted to better understand these orientations. It is also clear that more knowledge is required to understand the suit and helmet responses to dynamic loads, and to determine the interaction with the seat and restraints. This issue is somewhat unique to NASA and very little research has been conducted to directly address these issues.

In addition, investigations of human tolerance of dynamic loads have been primarily conducted on young, healthy males, or elderly male PMHS. The role of gender, age, and anthropometry on injury risk has been addressed to varying degrees in the past, but more research is needed to understand the effect of gender on injury risk in the spaceflight context, particularly coupled with the suit and helmet. Finally, spaceflight deconditioning is a risk factor unique to spaceflight, and requires additional research to better understand the effect of this factor.

Although injury assessment methods have improved dramatically over the past 5 decades, there is still no single method that satisfactorily addresses all of the risk factors and other considerations. In addition, the prediction of the very low injury risks associated with dynamic loads requires additional research. The available numerical models have all been developed for other environments and additional research is required to adapt or validate these models for spaceflight injury prediction.

Knowledge Gaps:

- OP1: Quantification of the risk of injury due to vehicle orientations and complex dynamics
- OP2: Quantification of the risk of injury related to the suit and helmet, particularly in relation to the seat, restraints, and crewmember anthropometry
- OP3: Quantification of the risk of injury related to gender, age and anthropometry
- OP4: Quantification of the risk of injury due to spaceflight deconditioning
- OP5: Determination of criteria for low injury risk (<5%)
- OP6: Adequate assessment methods validated for the spaceflight environment

7.0 CONCLUSION

During spaceflight, crewmembers are exposed to dynamic loads which have the potential to cause injury. Dynamic loads are transient loads (≤500ms) which are most likely during launch, launch or pad abort, and landing.

Several extrinsic factors affect the risk of injury including: vehicle dynamic profile, the seat and restraint design, and the spacesuit and helmet designs. Because each vehicle can have different launch, abort and landing dynamics, the risk of injury is greatly influenced by the vehicle design. Vehicles which minimize crew exposure to dynamic loads will be inherently safer than vehicles which have higher dynamic loads. The seat and restraint designs also contribute to the risk of injury (or mitigation) depending on the design and how effective they are at minimizing movement of the human. Finally, the spacesuit and helmet may contribute to the risk of injury if not properly designed. The suit can hinder the effectiveness of the restraints on the crewmember. Any rigid elements or the helmet can impart point loading and cause blunt impact. The helmet can pose a risk due to the mass of the helmet if it is not properly supported.

In addition to these extrinsic factors, there are additional intrinsic factors of the crew that can contribute to the risk of injury. These are: age, gender, anthropometry, and deconditioning due to spaceflight. Age has been shown to be a risk factor in other analogous environments such as automobile collisions. Gender can also influence injury risk, as body strength and geometry can differ between men and women. Anthropometry has been found to have an effect on injury risk since loads may not be proportional to the difference in anatomical structure as well as strength. Finally, spaceflight deconditioning has been shown to cause decrements in bone mineral density and muscle strength, which could affect the crewmember's tolerance to dynamic loads.

To assess injury risk, there are multiple methods available, although each have advantages and disadvantages. The methods can be divided into 3 categories: humans, human surrogates, and numerical models. Although human data seem to be the ideal solution for assessing injury risk, there are several drawbacks. Human volunteer testing is limited to sub-injurious levels but allows subjective feedback. Post-mortem human subjects (PMHS) can be tested at injurious levels, but cannot be used to investigate living tissue responses to trauma and do not include active muscle tone. Human exposure data contains cases of living human injury, but do not allow for prospective investigations of injury mechanisms. Human surrogates include Anthropomorphic Test Devices (ATD) and animal models. ATDs are not biofidelic in all instances and are not able to predict injury in all conditions; however, they are easily tested and have reproducible data. Animal models allow prospective testing of living tissue, but are not anatomically identical to humans. In addition, numerical models are available to assess injury risk. Dynamic response models are simple, but are limited in their injury prediction capabilities. ATD Finite Element (FE) models have similar limitations as the physical ATDs. Human FE models have great potential for allowing injury predictions; however, currently they are not validated in all necessary conditions. Finally, regardless of the method used to assess injury risk, adequate criteria for assessing low risk of injury (<5%) are needed.

Given this evidence, multiple knowledge gaps still exist in our understanding of the risk of injury to dynamic loads. These gaps include: the effect of various body orientations on injury risk during spaceflight; the effect of suit, seat and restraint designs on injury risk; the effects of the age, gender and anthropometry on injury risk; the effects of spaceflight deconditioning on injury risk; criteria to adequately assess low risks of injury; and adequate methods for assessing injury risk. These knowledge gaps highlight areas of needed research to assist in mitigating the risk.

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10.0 LIST OF ACRONYMS

ACES	Advanced Crew Escape System
AIS	Abbreviated Injury Scale
ATD	Anthropomorphic Test Device
BDR	Brinkley Dynamic Response
DXA	Dual Energy X-Ray Absorptiometry
EVA	Extravehicular Activity
FAA	Federal Aviation Administration
FARS	Fatality Analysis Reporting System
FE	Finite Element
GHBMC	Global Human Body Model Consortium
IRB	Institutional Review Board
IVA	IntraVehicle Activity
L/D	Lift to Drag Ratio
LOC	Loss of Crew
LSTC	Livermore Software Technology Corporation
MADYMO	MAthematical DYnamic MOdel
NASA	National Aeronautics and Space Administration
NHTSA	National Highway Traffic Safety Administration
ORIS	Operationally-Relevant Injury Scale
OSU	Ohio State University
Ph.D.	Doctor of Philosophy
PMHS	Post-Mortem Human Subjects
SID	Side Impact Dummy
SMC	Suit Mounted Connector
THUMS	Total HUman Model for Safety
US	United States
USSR	Union of Soviet Socialist Republics
V*C	Viscous Criterion
WPAFB	Wright-Patterson Air Force Base

11.0 ADDITIONAL REFERENCES

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